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Second-Generation Hand-Held Force Magnifier for Surgical Instruments

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Abstract. We have developed a novel and relatively simple method for magnifying forces perceived by an operator using a surgical tool. A sensor measures force between the tip of a tool and its handle, and a proportionally greater force is created by an actuator between the handle and a brace attached to the operator's hand, providing an enhanced perception of forces at the tip of the tool. Magnifying forces in this manner may provide an improved ability to perform delicate surgical procedures. The device is completely hand-held and can thus be easily manipulated to a wide variety of locations and orientations. We have previously developed a prototype capable of amplifying forces only in the push direction, and which had a number of other limiting factors. We now present a second-generation device, capable of both push and pull, and describe some of the engineering concerns in its design and our future directions.

Keywords: haptics, touch, robotic surgery, microsurgery, force magnifier, force-reflecting, steady hand

1 Introduction

A need exists for improvement in the perception of forces by the sense of touch when using tools to perform delicate procedures. One key area for potential applications is ophthalmological surgery, in which we have recently been exploring techniques for image-guided intervention using optical coherence tomography [1]. Therefore, this area in particular has motivated us in the present research. The technology we are developing could, however, also prove helpful in other forms of microsurgery. For example, surgeons routinely repair tiny blood vessels under a microscope that are far too delicate to be felt by the hand of the surgeon. Providing a useful sense of touch for such applications could improve outcome and increase safety in any of these applications.

Purely telerobotic systems such as the da Vinci® Surgical System (Intuitive Surgical, Inc.) can provide motion-scaling, so that fine motion of the tool can be controlled by coarser motion of the operator's hand on the controls. Although force at the tool tip cannot be sensed by the operator in the current commercial da Vinci® device, experimental systems have been tested that translate these forces into visual cues [2] as well as into vibrotactile feedback to the operators fingers [3].

A different, non-telesurgical approach has been demonstrated in several experimental systems, including the Force-Reflecting Motion-Scaling System created by Salcudean, et al. [4] [5], and the Steady Hand Robot described by Taylor, et al. [6][7]. These generate a magnified sense of touch by using a robotic arm that holds the surgical tool simultaneously with the surgeon, pushing and pulling as appropriate, to amplify forces detected by small sensors between the handle of the tool and its tip. Because every force requires an opposing force, the robotic arm must be mounted somewhere, and its weight must be supported by that mounting. Thus in these systems the magnified forces are created between the tool handle and subsequently the floor. To permit free motion of the tool by the surgeon, an elaborate remote-center-of-motion articulated robot arm is required, along with a control system to keep the tool moving naturally, as if controlled just by the operator, so that the surgeon can have something approaching the degrees of freedom and ease of manipulation that he/she is accustomed to with a hand-held tool. Such systems are typically fairly extensive and complex. Issues arising from the limited and congested workspace typical in microsurgery raise serious challenges to their practical deployment.

The desire to free robotic surgery devices from the floor-standing robotic arm has led to hand-held systems such as the Micron microsurgical instrument from Riviere's group, which uses piezoelectric actuators to move the tip relative to the handle, based on optical tracking of both the tip and handle [8]. The primary goal of Micron is to reduce the effects of hand tremor. It is not suited to provide a magnified sense of touch.

Another hand-held probe is the MicroTactus, developed to enhance tactile sensitivity during minimally invasive surgical tasks such as probing and exploration [9]. The device is instrumented with an accelerometer attached to the tool tip, as well as a solenoid operating inertially against an internal movable weight in the handle to create vibrotactile stimulation. The accelerometer is oriented orthogonal to the actuator to decouple the input signal from the output. Since the forces are generated purely by the inertia, they are inherently transitory and must integrate to zero over time. The device is therefore capable only of communicating texture while moving across a surface, rather than non-transitory forces such as those from sustained pushing and pulling against a target.

When the goal is to magnify non-transitory forces for the operator to feel, some external frame to "push against" has generally been required. The field of haptic simulation faces the same dilemma of generating sustained forces for the fingers to feel without anchoring the renderer to some solid base. Recent examples of such portable solutions include the "active thimble" described by Solazzi, et al. [10]. The device is entirely mounted on one hand. It attaches to the proximal part of the finger and reaches over to contact the fingertip, thus generating forces between two parts of the operator's own anatomy. As they describe it, "[a] limit of traditional kinesthetic interfaces is the difficulty to achieve a large workspace without a detriment of dynamic performance and transparency or without increasing the mechanical complexity. A possible solution to overcome this problem is to develop portable ungrounded devices that can display forces to the user hands or fingers."

In our approach we extend the concept of ungrounded haptic devices from purely virtual environments to real tools, by including a force sensor for interaction with an actual target. As in the Force-Reflecting and Steady Hand systems described above, we provide a magnified perception via the tool handle of forces sensed at the tool tip. Our approach, however, does not require any freestanding apparatus, and instead produces forces between portions of the operator's own anatomy. The concept is shown in Fig. 1. A hand-held tool contains a sensor, which measures force f between the handle and the tip. This signal is amplified to produce a force F in the same direction on the handle of the tool, using an actuator mounted on the back of the hand (in this case, a solenoid). The human is, in effect, providing a moving platform from which the magnified forces are generated.

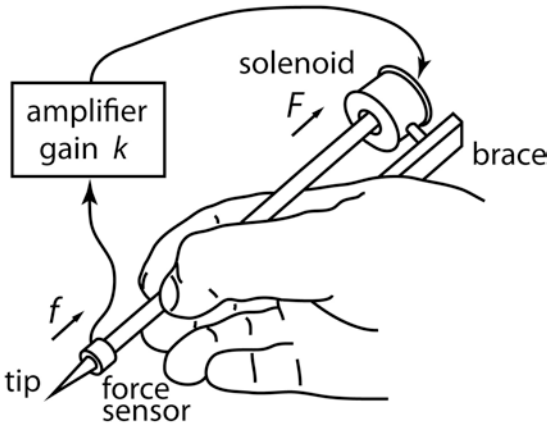

Fig. 1 The Hand-Held Force Magnifier (HHFM) uses a sensor to measure force f between the handle and the tip, which is amplified to produce a force $F = k\,f$ in the same direction on the handle using a solenoid mounted on the back of the hand.

2 Model-1 Hand Held Force Magnifier

We have previously reported on the Model-1 prototype of the Hand-Held Force Magnifier (HHFM) [11], illustrated in Fig. 2. We briefly review it here. In this initial prototype, the small button on a force sensor (Honeywell FS01, 0-6.7 N) served as the tool tip. The tool handle was the body of a syringe attached to a piece of 1/4 inch brass tubing containing a stack of 8 permanent rare-earth magnets (3/16" Radio Shack 64-1895) inserted into a custom solenoid (250 ft of 30 gauge wire, 25 ohms, approx. 2360 turns). The solenoid was attached by a dual gimbal to a brace, which was mounted to the back of a wrist splint strapped to the operator's right hand. The dual gimbal permitted free rotation in azimuth and altitude, while maintaining a relatively tight connection for the transmission of force in the range (axial) direction. A control system (not shown) consisted of a linear amplifier capable of supplying 32 V at 2 A, enough to operate the solenoid over its maximum range, to produce a force F from the solenoid of up to 1 N, proportional to, and in the same direction as, the force f sensed at the tool tip (Figs. 1 and 2). Thus we can express the system's behavior as simply

$$F = kf. \tag{1}$$

The proportionality factor *k* was adjustable from 0 to 5.8. Above this level, the system became unstable.

The Model-1 prototype is shown in Fig. 2 being used to push a spring from the side to bend it. With the gain *k* set to 0, the spring is felt through the tool to be subjectively quite easy to bend. With *k* increased to maximum, the spring feels much harder to bend. This haptic illusion is due to the fact that the fingertips must match not only the force of the spring *f* but also of that of the solenoid *F*. Thus the device magnifies the operator's sensation of touch while keeping the force actually applied to the spring relatively small.

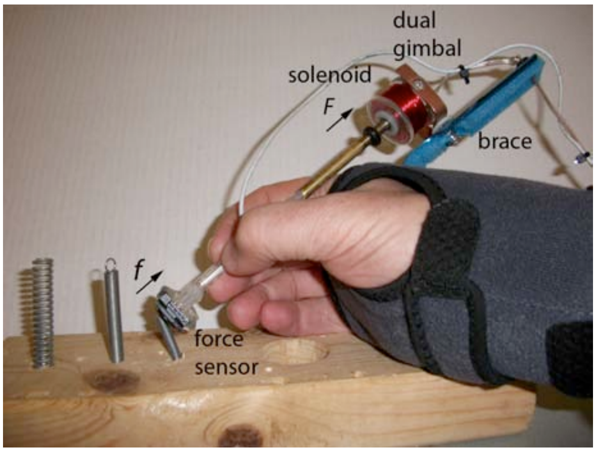

Fig. 2 Model-1 prototype of the Hand-Held Force Magnifier (HHFM). Here the operator "feels" a magnified force generated pushing against a spring.

We hypothesized that the operator can sense forces at the tool tip that are smaller than would otherwise be perceivable, and can control these smaller forces with greater delicacy by interacting with the magnified forces. Two experiments were conducted to characterize the Model-1 HHFM. In the first experiment, the participants' ability to sense the presence of a force (i.e. the force detection threshold) was measured with and without the use of the HHFM. In the second experiment, we used a method of magnitude estimation to characterize the impact of the device on the subjective force intensity. Both studies produced a measure of the perceptual magnification of force implemented by the HHFM, and clearly showed that percepts of force at both threshold and supra-threshold levels are rescaled when using the HHFM, demonstrating that the force magnification induced by the device is well perceived by human users [11].

Anecdotally, it was clear that the spring being pushed in Fig. 2 became subjectively stiffer when the HHFM was activated. Surprisingly, this was true even when the wrist was not resting on the bench top, when the operator was delivering force solely through the shoulder. This was unexpected, since all of the HHFM's magnifying effect is physically due to forces generated within the hand, and yet the clear perception to the operator was that the muscles of the arm and shoulder are encountering a stiffer spring. Clearly, some form of perceptual integration is occurring.

A number of shortcomings were evident in the Model-1 HHFM. Since the sensor was only able to register positive forces, only pushing interactions could be magnified. The wrist-splint used to mount the device on the hand was cumbersome, restrictive, and appropriate only for relatively large hands. The analog circuitry used to control the actuator was limited in terms of its ability to implement more complex control strategies than simple linear gain. Finally, the solenoid tended to overheat and the simple bearing was prone to binding.

3 Model-2 Hand-Held Force Magnifier

We report here on our recent complete redesign of the HHFM, as embodied in our Model-2 device.

In the Model-2 device we have replaced the wrist-splint with a streamlined and better fitting brace and actuator assembly (see Figs. 3 and 4) constructed of aluminum stock and acrylic tubing. The two-piece brace is secured to the hand by Velcro straps, permitting rapid attachment to, and detachment from, the hand. Foam padding on the back of the brace improves fit and comfort. A rotary bearing is press-fit into the top of the brace to allow movement of the actuator assembly in azimuth. A single hinge allows for movement in altitude. Instead of a solenoid, we used a commercially available voice coil (LVCM-19-022-02, Moticont, Inc.) as

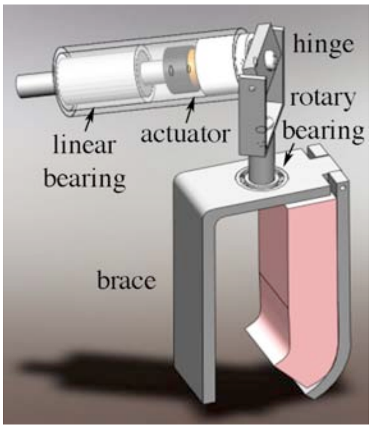

Fig. 3 Model-2 brace and actuator assembly.

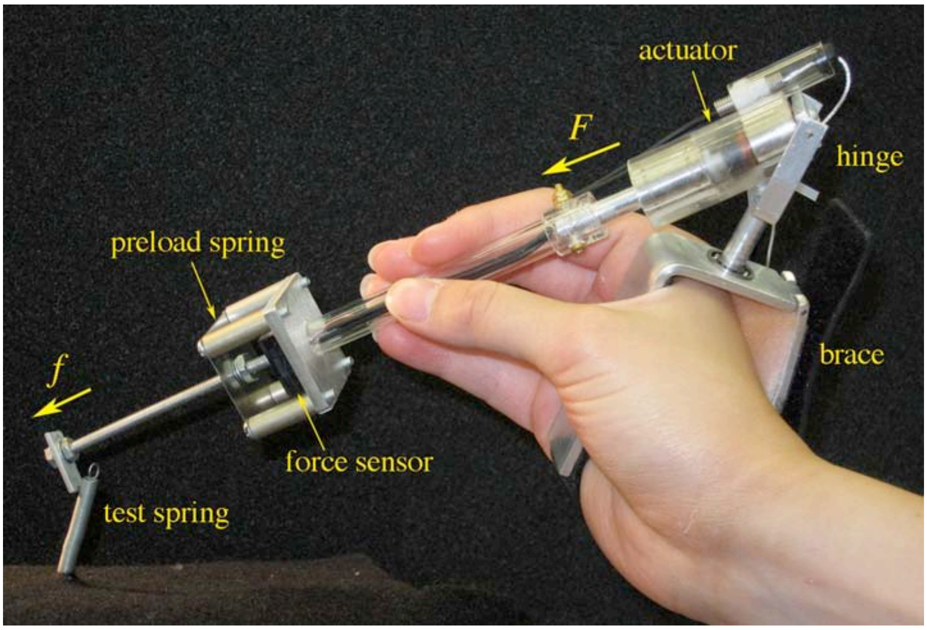

Fig. 4 Model-2 prototype of the Hand-Held Force Magnifier (HHFM), capable of both push and pull. Here the operator "feels" a magnified pull against a test spring.

an actuator, capable of generating up to 2.5 N of force in either direction over a 12.7 mm stroke length. In a voice coil, a coil of current-carrying wire moves inside a stationary magnetic housing, as opposed to a solenoid, which features a moving metallic or magnetic bar in the presence of a stationary coil. Thus a voice coil is capable of quicker response and is less sensitive to angular dependency on gravity. An aluminum post was secured to the voice coil and supported by a concentric linear bearing to allow the handle to move freely with minimal friction.

A major goal of the Model-2 was to make it bidirectional, able to amplify both push and pull. A small lateral bar was added to the tool tip to permit both pushing and pulling forces to be applied (see Fig. 4). Notice that the direction of arrows for both the detected force f and the amplified force F are reversed in Fig. 4 from Fig. 2, indicating that Fig. 4 depicts an amplified sense of pulling on the test spring. This bi-directionality was accomplished using the same push-only force sensor used in the Model-1 by adding a preload spring made of steel shim-stock. The spring produced a steady force that could be added to (push) or relieved (pull).

Another advance in the Model-2 was the incorporation of a microprocessor (ADUC0726, by Analog Devices). The 40 MHz processor has multiple 1 MS/s, 12-bit A/D and D/A convertors. With it, we are able to run a number of fairly complex algorithms written in the C programming language, maintaining a 10 kHz through-put without significant jitter. This capability has permitted us to solve the problem of zeroing the preload voltage (the "tare" switch on a typical scale for weighing) by averaging a random sample of the sensor voltage over a period of inactivity. The microprocessor furthermore permits rapid exploration of a whole range of more complex functions beyond simple gain, including non-linearity, time varying behavior, and Proportional-Integral-Differential (PID) control to increase stability.

We are currently testing the Model-2 HHFM using a Magnetic Levitation Haptic Device (MLHD) (Maglev 200TM from Butterfly Haptics) to accurately control forces and displacements (see Fig 5). The MLHD uses Lorentz forces for actuation, which arise from the electromagnetic interaction between current-carrying coils and magnets. It was developed by co-author Ralph Hollis [12]. Since there are no motors, gears, bearings, or linkages present, the MLHD is free of static friction and able to generate forces precisely with a resolution of 0.02 N. We previously used the MLHD to study simple force magnification with the Model-1 HHFM, and we are now testing the Model-2 using more complex simulations of surgical tasks, such as piercing a sheath of connective tissue or peeling a membrane.

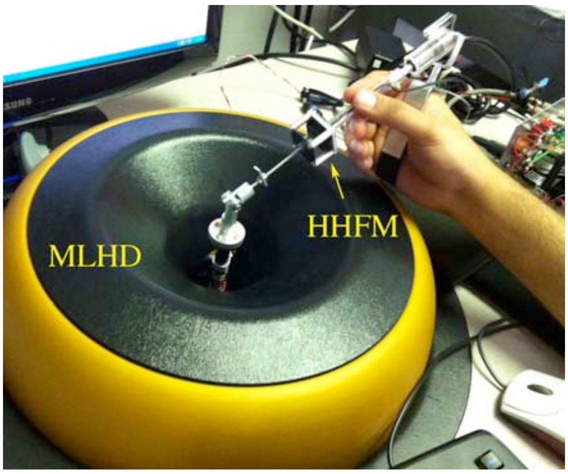

Fig. 5 Magnetic Levitation Haptic Device (MLHD) serving as test bed for the Model-2 Hand-Held Force Magnifier (HHFM).

4 Discussion

When we initially developed the HHFM, we were considering the clinical need for greater sensitivity to forces during eye surgery, but many applications may actually be suited for this technology. We have listed some of these in Table 1. Most involve rigid tools, such as needles, scrapers, hooks, scalpels and blunt dissectors used in microsurgery, where forces may be so delicate as to be impossible to feel. However, we have come to understand that such minute forces may be present in regular surgery as well, especially with sharp tools designed, after all, to minimize the forces resisting cutting and stabbing. Furthermore, potential uses of the HHFM may include operating in the "bloody field" of cardiac surgery, where a magnified sense of touch may improve the surgeon's ability to feel structures when vision is obscured by blood. The technology may also be adapted for use at the end of a catheter, permitting axial force and axial torque to be felt from the handle of the device, for example, to navigate the branch points in a vein or bronchus.

Table 1

Eye Surgery

- Cataract surgery
- Vitrectomy
- Surgery for Glaucoma
- Retinal surgery
- Corneal surgery
- Retrobulbar Injection
- Intra-ocular tumor excision

Other Microsurgery

- Neurosurgery
- Vascular & Microvascular Surgery
- Plastic Surgery
- Trauma and Reconstructive Surgery

Other Surgical Applications

- Epidural needle insertion
- Jugular vein needle insertion
- Cardiac surgery (bloody field)
- Minimally Invasive Surgery (laparoscopic, cardiac)
- Catheter based vascular procedures

The question of whether to ampligy non-axial forces and torques (those not along the axis of the tool) also arises. Certainly, lateral forces at the tip of a rigid tool do exist and are partially converted to torques at the handle. The resulting mechanical disadvantage of effectively operating from the short end of a lever makes amplification of such torques inefficient. Another way to look at this is that non-axial forces are inherently already magnified, in the sense that such forces are only exerted with significant mechanical disadvantage.

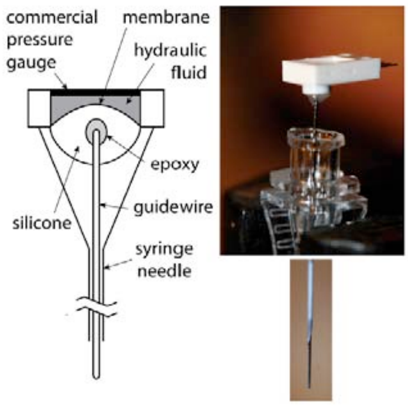

Fig. 6 Guidewire pressure sensor prototype.

Sterility will be a concern in most surgical applications. We envision building the brace and actuator into a surgical glove in a way that various portions may be detached, some being reusable and sterilizable, while others are pre-sterilized and disposable. In particular, removable disposable tips would be a practical solution to the need for a variety of force-magnified tools during a given procedure.

The choice and location of the force sensor is a particularly important aspect in the design of a clinically practical HHFM. The force sensor used in the Model-1 and Model-2 is clearly not ideal, because of its large size. It was chosen for convenience because it comes pre-calibrated and temperature compensated. Smaller sensors are available with some extra effort. More central to the design is the question of optimal sensor location. One possibility is to locate the sensor more proximally, in the handle, or behind the handle, and to communicate with the distal tip by means of a mechanical or hydraulic linkage. We have explored the possibility of a proximal force sensor, with only limited success thus far, using a commercially available pressure sensor commonly employed in catheters (Motorola MPX2011DT1). This sensor was chosen for its small size (6.60 mm × 6.07 mm × 3.81 mm), high sensitivity (full scale pressure limit of 75 kPa), and low cost (less than $1). As shown in Fig. 6, the sensor consists of a piezoresistive strain gauge with a pocket of hydraulic fluid held against its forward face by a very thin membrane. The back face of the sensor is exposed to atmospheric pressure, allowing for both positive and negative pressures to be measured relative to the atmosphere. We first explored attaching this sensor to a syringe and transmitting pressure to it via a tiny plunger in the tip of the syringe needle. This proved ineffective, in that the design of the distal plunger was problematic. We then used a mechanical linkage consisting of a guidewire within the shaft of the syringe, attached to the membrane of the pressure sensor by means of a small epoxy droplet within a flexible silicone mantle (again see Fig. 6). The pressure sensor requires pre-amplification, which we accomplished using an operational amplifier mounted just behind the sensor in the handle for optimal signal-to-noise. The entire apparatus, attached to the Model-2 brace and actuator assembly, is shown in Fig. 7. This overall design proved quite sensitive to both pushing and pulling, but suffered from significant hysteresis due to friction of the guidewire against the inside of the needle.

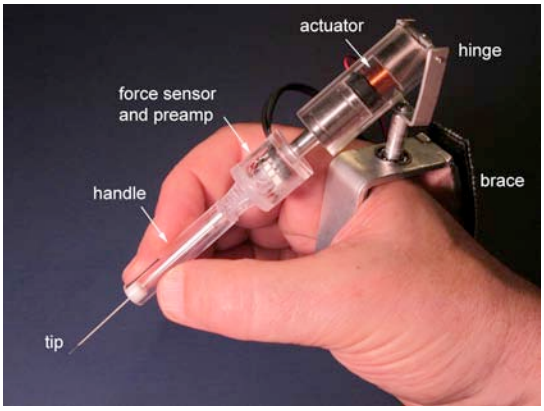

Fig. 7 HHFM design with proximal pressure sensor and guidewire linkage.

Another approach we are considering is to keep the force sensor distal to the handle but to greatly reduce its size. Very small piezoresistive surface mount pressure sensors are available (for example, 2.5 mm x 3.3 mm x 1.3 mm,

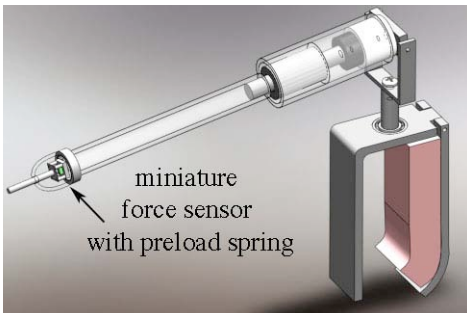

Fig. 8 HHFM design with miniature distal force sensor.

3000 series from Merit), which could be enclosed in the distal handle, as shown in Fig. 8. Moving closer to the tip to measure the force reduces interactions between orthogonal forces and torques. These may also be disambiguated by micro-machined arrays of strain gauges, such as those developed by Berkelman, et al. [14]. For distal sensing, an appealing solution is the optical Bragg sensor, which is small enough to be built into a fine needle tip, and which may be interrogated via optical fiber. One such system is being developed by Sun, et al., for use in retinal surgery, in which having the sensor in the tip eliminates confounding forces resulting from insertion through the sclera [15].

Finally, an ongoing concern is the actuator. Voice coils, as well as conventional solenoids, produce forces that vary with translation of the ferromagnetic element relative to the coil. For this reason, producing known forces requires either measuring that translation or limiting it to a very small range. Depending on whether pushing or pulling forces are being actuated, one or the other range boundary must be avoided to permit the operator to sense the resulting actuated forces. We are considering various methods of accomplishing this. We are also considering other types of actuators with larger, more uniform, ranges of operation.

5 Conclusion

We have reported here on progress on the HHFM, including the design construction of a new working prototype. We have also discussed the progress in our thinking about future evolution of the device and its likely applications in clinical medicine. The major contribution of our work, we believe, is to provide a magnified sense of touch without requiring an external robotic arm. The force that was generated between the operator's hand and the floor by the robotic arm in previous implementations of force magnification is replaced by a force generated between two locations on the operator's hand, freeing the design to permit a small, light, hand-held device.

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